BASIC PHYSICAL PRINCIPLES OF ULTRASOUND

INTRODUCTION

C ompetent performance of an ultrasound examination of any type is reliant on an understanding of the basic physical principles that govern the production of ultrasound waves and their interaction with biological tissues of differing densities and consistencies. It is also necessary to understand the basic concepts of how the ultrasound systems work and the concepts underlying their manipulation of echoes.

Given a working knowledge of these various concepts and an understanding of how the theoretical principles can be put to use in a practical arena, physics of ultrasound becomes the bedrock upon which to build all the other knowledge necessary in becoming a competent sonographic diagnostician.

Lectures and notes provided during this course are the barest essentials of the learning you will require and our emphasis will be on the practical application of the principles involved. Further theoretical knowledge can be gained by wider reading and study. Recommended texts are as follows:

1. Applied Physics and Technology of Diagnostic Ultrasound; Author . Roger Gent
2. Diagnostic Ultrasound, Principles and Instruments (Fifth Edition); Author . Frederick Kremkau

WHAT IS ULTRASOUND?

Simply stated, ultrasound is sound whose frequency is above the range of human hearing. Diagnostic ultrasound is used to evaluate a patients internal organs. Sound waves are transmitted into the body; then, because the various internal structures reflect and scatter sound differently, returning echoes can be collected and used to form an image of a structure.

Sound waves consist of mechanical variations containing condensations or compressions (zones of high pressure) and rarefactions (zones of low pressure) that are transmitted through a medium.

Beam Direction
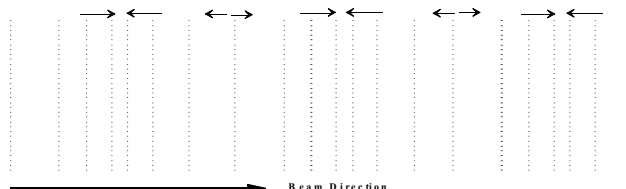
Unlike X-Rays, sound is not electromagnetic. Matter must be present for sound to travel, which explains why sound cannot propagate through a vacuum. Propagation of sound is the transfer of energy - NOT MATTER . from one place to another within a medium, some energy is also imparted to the medium.

Sound is categorised according to its frequency (number of mechanical variations occurring per unit time):

Human Audible range   20 . 20000Hz
Diagnostic Ultrasound   1 . 20MHz
                        (3-12MHz being the most common range)

WAVE TERMINOLOGY

The characteristics of a sound wave can be described by the following parameters:

- Period (T) . the time taken for a particle in the medium through which the wave is traveling to make one complete oscillation about its rest position. (One oscillation is also referred to as a cycle)

- Frequency (f) . the number of cycles per second performed by the particles of the medium in response to a wave passing through it. Expressed in Hertz, where I Hz = 1 cycle passing a given point each second, therefore 3MHz = 3 million cycles per second

- Wavelength (λ) . the distance between two consecutive, identical positions in the pressure wave (e.g. between 2 compressions or between 2 rarefactions). It is determined by the frequency of the wave and the speed of propagation in the medium through which it is traveling. In diagnostic ultrasound, commonly used frequencies and their respective wavelengths in soft tissue are as follows:
Frequency - 3MHz / Wavelength . 0.51mm
Frequency . 5MHz / Wavelength . 0.31mm

- Velocity (c) . speed of sound with direction specified. When a sound wave travels through any medium it is certain parameters of that medium, which determine the speed of sound propagation. These determining factors are density and compressibility. Therefore, speed of sound is a characteristic of each material through which sound travels; e.g.

Material	Speed of Sound
Air	330 metres/second
Metal	5000 m/sec
Pure Water	1430 m/sec

  Speed of sound in various organs of the body also differs; e.g.

 | Soft Tissue Type | Speed of Sound |
  | --- | --- |
  | Fat | 1450m/sec |
  | Liver | 1550m/sec |
  | Blood | 1570m/sec |
  | Muscle | 1585m/sec |
  | Bone | 4080m/sec |

In practical terms, ultrasound machines need to operate on a single value for speed of sound in soft tissue; therefore an average speed for soft tissue is taken to be **1540m/sec**.
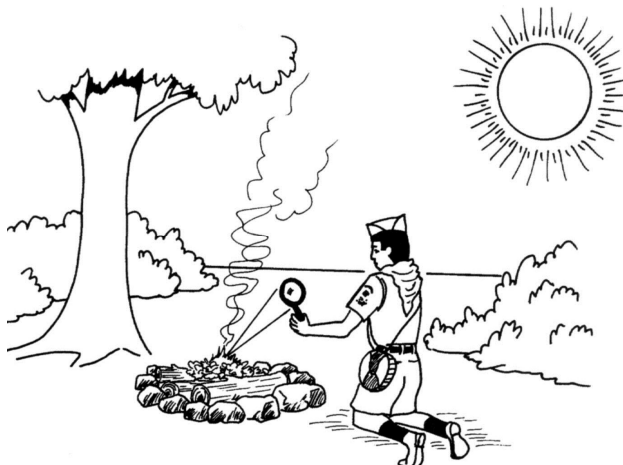
**REMEMBER!!
Speed of Sound in Soft Tissue
= 1540m/sec**

➲ *This can be extremely important when learning about artifacts later, since differences between the actual speeds of sound and those inferred by the ultrasound machine can cause misregistration of echoes on the image.*

- Amplitude (A) . maximum variation of an acoustic variable, thus amplitude is a measure of the degree of change within a medium when a sound wave passes through it and relates to the severity of the disturbance. In this way, the amount of energy in a sound wave can be determined. Amplitude is expressed in units that are appropriate for the acoustic variable considered.

Amplitude

Pressure
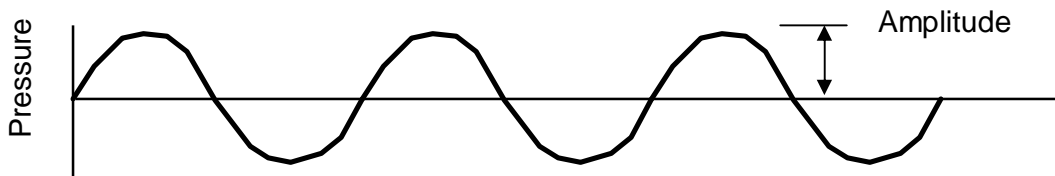
- Power (W) . the rate at which work is done or the rate of flow of energy through a given area. In diagnostic ultrasound energy is contained within the beam, so the power is the rate of flow of energy through the cross-sectional area of the beam. Power is expressed in Watts.

- Intensity (I) . power per unit area. Intensity is expressed in milliwatts per square centimeter ($mWatts/cm^2$). Intensity is an important parameter in describing an ultrasound beam and in the understanding of bioeffects and safety. The example of sunlight shining on wood shavings is often used to illustrate this phenomenon . Sunlight does not usually burn wood shavings but when the sunlight is concentrated into a small area (increased in intensity) by a magnifying glass, the wood shavings can be burnt. So, power remains the same, intensity is increased and an effect is produced (increased heat).

The scout can burn the wood shavings with the aid of a magnifying glass to focus the sun's rays, therefore increasing the intensity.
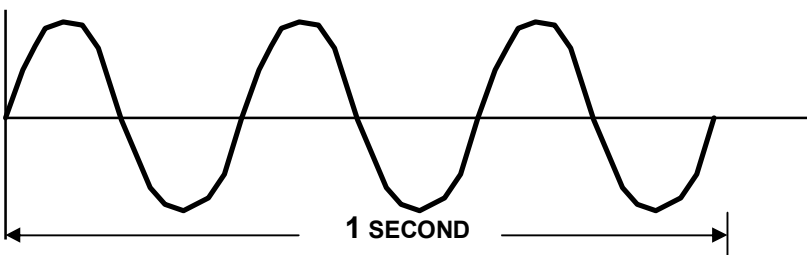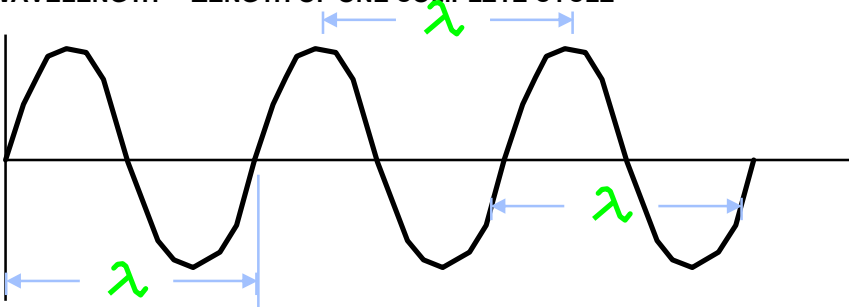
➲**IMPORTANT SUMMARY DIAGRAMS TO REMEMBER**

CYCLE – ONE COMPLETE OSCILLATION OF THE WAVE
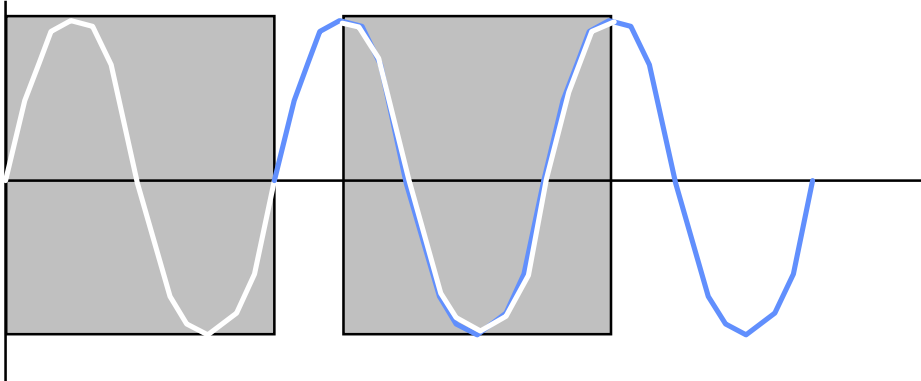
FREQUENCY – NUMBER OF CYCLES PER SECOND OF TIME

1 SECOND

**WAVELENGTH – LENGTH OF ONE COMPLETE CYCLE**

λ

λ

λ

Frequency and wavelength have an inverse relationship shown in the formula $c = f\lambda$, where c is the speed of sound in tissue, f is the frequency and λ is the wavelength. Therefore, the higher the frequency, the shorter the wavelength. This relationship is fundamental to practical scanning since it has a considerable effect on the achievable penetration depth and resolution of the ultrasound beam.

THE PIEZO-ELECTRIC EFFECT

Artificially grown crystals are commonly used for modern transducers and are treated with high temperatures and strong electric fields to produce the piezo-electric properties necessary to generate sound waves

The principle of piezo-electricity is central to the production of ultrasound beams and states that some materials produce a voltage when deformed by an applied pressure and produce a pressure when deformed by an applied voltage.

Thus, when a voltage is applied to the faces of a crystal, it expands or contracts depending upon the polarity of the voltage applied. The crystal then resonates, converting electricity to ultrasound. The frequency of sound produced is dependant on the thickness of the crystal. Conversely, the when the crystal receives an echo, the sound deforms the crystal and a voltage is produced on its faces . this voltage is then analysed by the system.
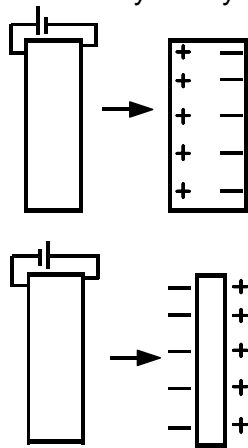
**Piezo-electric effect: Reversing polarity of voltage
Crystal undergoes expansion & contraction**

If expansion & contraction occurs more than 20000 times per second, then ultrasound is being produced, which will continue until the applied voltage is discontinued (Continuous wave). If voltage is applied for an extremely short time, then the crystal resonates (rings) at its own frequency and the ringing gradually decays . a pulse is produced. The larger the voltage applied the greater the amplitude of the emitted sound (louder ringing).

Very short pulses are required for diagnostic ultrasound images to be produced, so the ringing of the crystal is stopped short by a damping material being applied to absorb vibrations (a hand placed on a bell will stop the ringing).

## PULSE-ECHO IMAGING

Images produced by diagnostic ultrasound systems are formed by sending out a short pulse of sound along a narrow beam path and waiting for return echoes to be collected from structures encountered along that path. The instrument generating and receiving those pulses of sound is the transducer, which is a critical component of the system due to its profound effect on image quality.

BASIC TRANSDUCERS

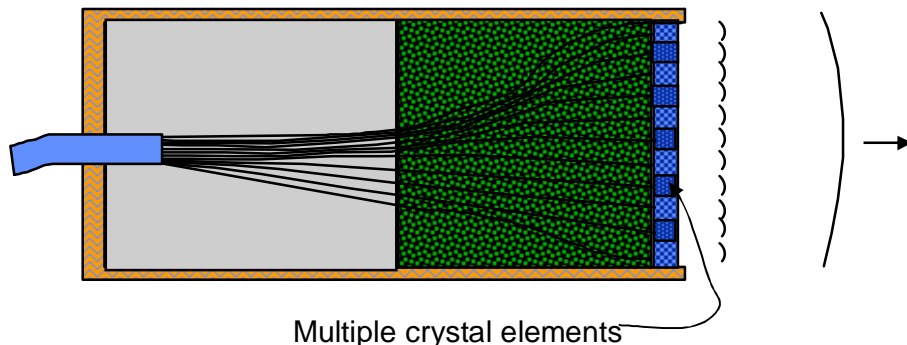Simply stated the basic transducer consists of three parts:

1. The Case
2. The Crystal
3. The Damping material

These are connected to the ultrasound system by electrical wiring.
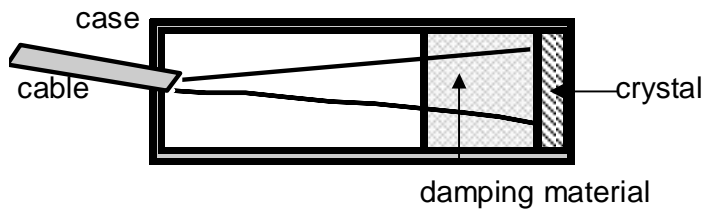
Multi-element Transducers

Most real time transducers use many small crystal elements for the formation of each pulse.

Multiple crystal elements

These form a line of elements and the transducers are known as linear array transducers. The wave fronts from each element combine to form a single wave front; this is known as Huygens Principle. Each individual element is acoustically and electrically isolated, allowing flexibility in beam formation.

BEAM PROFILE

The shape of the ultrasound beam is important to the quality of the image it produces. The shape of the beam from a simple transducer is shown below.

The beam profile is made up of three parts:
1. Near Field (Fresnel Zone)
2. Far Field (Frauhofer Zone)
3. Transition Point . point at which near field ends and divergence begins

(The length of the near field is often termed the transition distance)
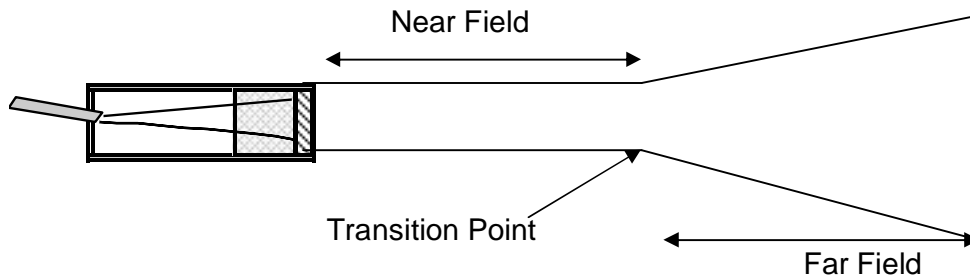
➲ **All sound beams are three-dimensional**

The near field is the part of the beam useful for imaging purposes; however this can be quite large in area, depending on the diameter of the crystal. Imaging requires a very narrow beam to produce high-resolution diagnosis. Focussing is used to achieve these narrow beams, there being various focussing methods in use throughout the manufacturing industry and new methods are being devised with each passing year.

Simple methods are to curve the crystal, add a curved lens in front of the crystal or focus by electronic means (the latter being the most common in today's transducers).

DIFFRACTION PATTERN

A beam profile of a simple transducer obtained by plotting the maximum pressure amplitude appears as in the diagram below and shows that the energy is not confined to a single lobe, but radiates off at various angles to the transducer face as 'off-axis' energy. These off-axis areas are called side-lobes.
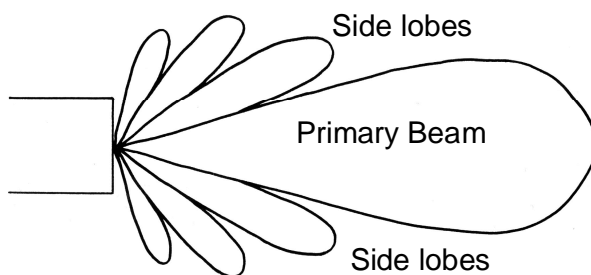
➲ **SIDE LOBES ARE THREE-DIMENSIONAL & SURROUND THE PRIMARY BEAM**

Array transducers also generate grating lobes which are defined as additional, weaker beams of sound energy travelling out in different directions from the primary beam as a result of the multi-element structure of transducer arrays.
Both side lobes and grating lobes are minimised as far as possible by the manufacturer during the design process but nevertheless still produce artifactual echoes within images. (See artifact section)
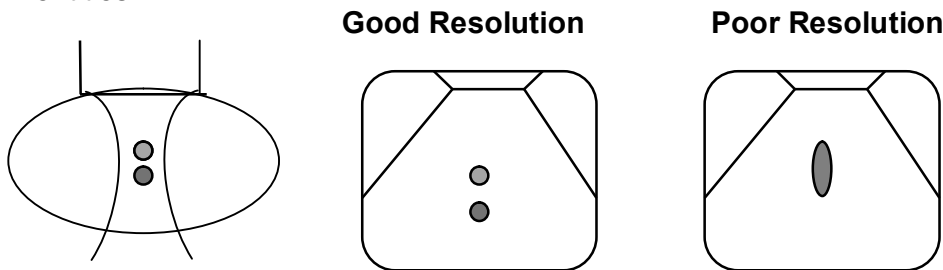
RESOLUTION

Resolution is defined, as the ability to distinguish echoes in terms of space, time or strength and good resolution is thus critical to the production of high quality images.

1. **CONTRAST RESOLUTION** refers to the ability of an ultrasound system to demonstrate differentiation between tissues having different characteristics e.g. liver/spleen.

2. **TEMPORAL RESOLUTION** is the ability of an ultrasound system to accurately show changes in the underlying anatomy over time, this is particularly important in echocardiography.

3. **SPATIAL RESOLUTION** is the ability of the ultrasound system to detect and display structures that are close together. Since an ultrasound image displays depth into the patient and width across a section of anatomy it is therefore reasonable to consider two types of spatial resolution . Axial & Lateral.

AXIAL RESOLUTION
The ability to display small targets along the path of the beam as separate entities.

Good Resolution   **Poor Resolution**

Axial resolution is dependent upon various factors, the most important of which being the length of the pulse used to form the beam. This is known as the spatial pulse length (SPL). The shorter the pulse length, the better the axial resolution. In fact the axial resolution limit is defined as being one half of the SPL.

⮑ **AXIAL RESOLUTION LIMIT = ½ SPATIAL PULSE LENGTH**

The SPL is determined by the number of cycles in one pulse and the length of each cycle (i.e. wavelength - this cannot be changed by the operator). However, wavelength, as we have already said, is inversely proportional to the frequency, so higher frequencies generate shorter wavelengths and therefore shorter pulse lengths. So we can say that higher frequency transducers will exhibit better axial resolution.

One note of caution should be that all transducers of a particular frequency will not necessarily have the same axial resolution, since other factors such as damping levels can have an effect on the SPL.

Other factors affecting the axial resolution, which the operator can control, are:

♦ Transmit power . the greater the amplitude of the voltage striking the crystal, the longer the ringing time and longer the SPL, leading to a slight decrease in axial resolution

♦ Received gain settings effect the length of the voltage signals generated by the returning echoes, the higher the gain . the poorer the axial resolution

♦ Field of view settings effect the display of pixels per unit area of the patient . smaller FOV setting makes best use of the available scan converter memory

LATERAL RESOLUTION

The ability to distinguish between two separate targets perpendicular to the beam path. (Same distance from the transducer)

Good Resolution   **Poor Resolution**
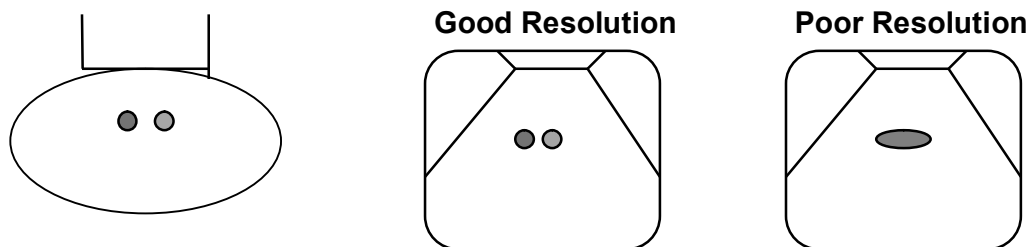
Lateral resolution is dependent upon the width of the ultrasound beam, that is . the wider the beam, the poorer the lateral resolution. Ultrasound machines assume that all received echoes arise from the central axis of the beam, so if two targets are within the beam at the same point in time, the echoes are assumed to have come from the same target and only one structure is registered on the image. If the beam is narrower than the distance between the two targets, only one target is within the beam at one time and as the transducer is swept across the body, both targets will be interrogated separately and both will be registered on the image. (See diagram)

DIRECTION OF SCAN   **IMAGE**

TARGETS SAME WIDTH APART
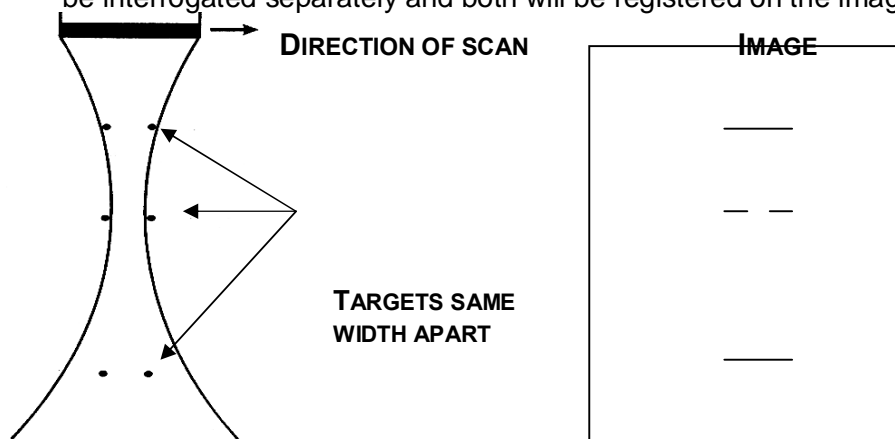
Where the beam width is at its narrowest, the targets are separated by a distance greater than the beam width and thus are resolved on the image.

As we can see, the lateral resolution is best where the beam is the narrowest and thus in the area of tightest focus in most transducers.

➡ **CORRECT POSITIONING OF THE FOCAL ZONES IS CRITICAL TO GAINING THE BEST LATERAL RESOLUTION FOR A GIVEN TRANSDUCER**

INTERACTION OF SOUND WITH TISSUE

ATTENUATION

As the sound beam traverses tissues within the body, various factors cause it to lose energy and therefore undergo a reduction in amplitude and intensity. This loss of energy is called attenuation. The amount of attenuation is determined by the tissue involved, the distance travelled and the frequency of the beam. The dependence of attenuation on frequency is of major importance in practical terms, since it is this dependence, which limits the penetration of an ultrasound beam of a given frequency and makes TGC controls a necessity in diagnostic ultrasound systems.

There are four main processes, which contribute to the attenuation of the sound beam, which are:
- **Reflection**
- **Refraction**
- **Absorption**
- **Scattering**

All of the above factors affect the echoes returning to the transducer as well as the transmitted beam.

> ↻ **The amount of attenuation depends on the total path length. (The time taken for the sound to reach the interface and the echoes to return)**

> ↻ **The approximate rate of attenuation in soft tissue is 1dB/MHz/cm**

REFLECTION

This occurs at interfaces between soft tissues of differing acoustic impedance, an alternative name for which is an acoustic impedance mismatch. Reflection occurs where the interface is large relative to the wavelength of the transmitted sound.

At tissue interfaces some of the sound is reflected and some is transmitted further into the soft tissue. The percentage of the sound reflected is dependent on the magnitude of the impedance mismatch and the angle of approach to the interface.

- **Soft tissue/air interface**   **- 99.9% reflected**
- **Soft tissue/bone**   **- 40% reflected**
- **Liver/Kidney**   **- 2% reflected**

Specular Reflections occur at large, smooth interfaces (e.g. Diaphragm, organ margins), in these cases the angle of incidence equals the angle of reflection.

Clinical Note:

> ↻ **Avoid bone, gas and air interfaces with soft tissue - little sound is transmitted which produces shadowing of deeper tissues.**

> ↻ **Try to use a soft tissue "window" to view deep structures**

REFRACTION

Refraction is the deviation in the path of a beam, occurring when the beam passes through interfaces between tissues of differing speeds of sound, when the angle of incidence to an interface is not 90°
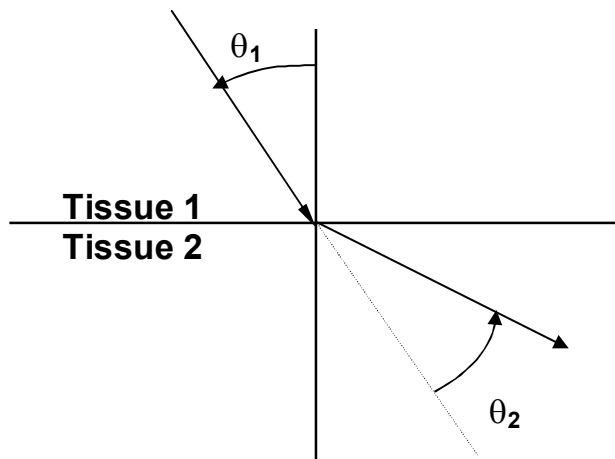
Refraction at an interface between tissues of differing velocities

Variations in velocity between soft tissue organs are generally fairly small (up to 10%). Deviations of the sound beam of up to 10° may occur, with resultant misregistration of echoes on the display.

➲**Refractions can produce some recognised artifacts.**

ABSORPTION

Transfer of some of the energy of the beam to the material through which the sound is traveling.

➲**Absorption increases with frequency**

➲**Absorption produces a heating effect**

➲**Clinical note:** **The increase in absorption with frequency explains why high frequency transducers cannot be used for examining deep structures within the body.**
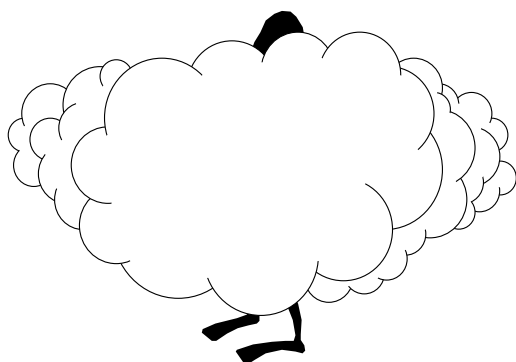
Understanding a little cloudy?

Absorption is a fairly complex subject, the physics of which we have not attempted to describe in detail.

Further reading from recommended physics text may blow away the clouds!

SCATTERING

Scattering occurs at interfaces within the sound beam path. The scatter pattern relies on the size of the interface relative to the wavelength of the sound.

- **Interface much larger than wavelength - Reflection (see page 10)**
- **Interface equal in size to wavelength – scatter in many directions with amount not being equal in all directions. (sometimes termed non-specular reflection)**

⮎**Clinical Note:**
Provides much of the textural information present in images and is dependent upon:

> **a. Frequency**
> **b. Angle of approach**

- **Interface much smaller than wavelength - Rayleigh scattering, which is equal in all directions (not angle dependent)**

⮎**Clinical Note:**
This type of scattering is caused by red blood cells and provides signals for Doppler assessment of blood flow.

IMAGE PRODUCTION

The images we use in diagnostic ultrasound are produced with B-Mode or brightness mode techniques. This can be described as a pattern of dots positioned on the monitor, as a result of the signals caused by returning echoes being processed by the system. The brightness of each dot is in proportion to the strength of the echo received.

As each pulse is emitted, its line of sight is known accurately and therefore the resulting echoes from sound/tissue interactions along this line are represented on the monitor in accurate spatial position.

Multiple lines of sight produced at known angles from the transducer then construct an image of the underlying tissue, composed of multiple thousands of dots. This image is known as a frame. Multiple frames produced in rapid succession form the moving or real-time image we see on the ultrasound monitor. Frame rates in modern systems are typically 10 . 30 frames per second and a flicker free image is achieved using electronic smoothing techniques.

TRANSDUCERS

Several types of transducer are available, their construction varies depending upon the type of examination they have been been produced to facilitate. Some of these are:
- Mechanical sector
- Electronic sector (phased array)
- Annular array
- Linear array
- Curved array

Specialised (intracavitary, intraoperative)

The types of transducer we use the vast majority of the time in general ultrasound practice are Linear and Curved arrays.

LINEAR ARRAY

The transducer is constructed with multiple small crystal elements arranged in a straight line (thus linear) across the face of the probe. These elements are pulsed sequentially in groups in order to produce a wider aperture and resultant improved beam shape.

The beams generated by these transducers travel at 90^0 to the transducer face. Thus the lines of sight are parallel, the line density is uniform from top to bottom of the image and the field of view is rectangular.
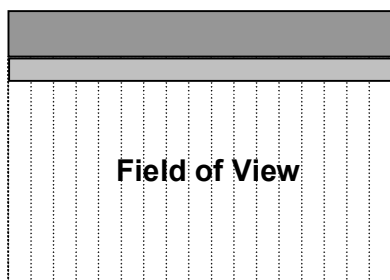
Field of View

CURVED ARRAY

SIMILAR IN CONSTRUCTION TO THE LINEAR ARRAY EXCEPTING THAT THE FACE OF THE TRANSDUCER IS CONVEX. SINCE THE BEAM PATHS REMAIN AT 90° TO THE TRANSDUCER FACE, THE RESULTING FIELD OF VIEW IS WIDER AT THE BOTTOM OF THE IMAGE WHILST REMAINING A REASONABLE WIDTH IN THE NEAR FIELD OF VIEW

Field of View
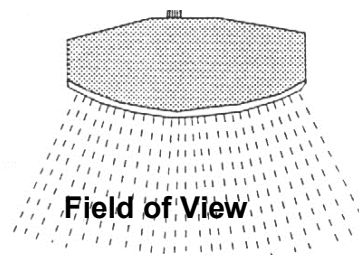
Transducers produced by different manufacturers may have similar frequency, size and shape, but can have quite different beam shapes and focussing due to the particular construction used by the manufacturer. They can have widely varying numbers of elements and so aperture widths.

➲ **Get to know your transducers and their capabilities**

ARTIFACTS

Ultrasound systems operate on the basis of certain assumptions relating to the interaction of the sound beam with soft tissue interfaces, such as:

1. A constant speed of sound in the body (assumed 1540m/s)
2. All echoes detected by the transducer originate from the central axis of the beam
3. The ultrasound beam travels in straight lines
4. The time taken for an echo from a given interfaces to return to the transducer is directly related to its distance from the transducer
5. The rate of attenuation of the beam is constant with depth and throughout the field of view

These assumptions are often incorrect, resulting in images that do not accurately reflect the anatomy of the scan plane. Several effects of these incorrect assumptions result in artifactual echoes with the display, these effects can produce:

1. Echoes in the display which do not appear in the correct position
2. Absence of echoes from parts of the image representing tissue which would normally generate echoes
3. Structures which are anatomically separate being displayed as joined
4. Brightness of echoes which may not correspond to that expected from a particular organ
5. Single structure being displayed as two separate structures
6. Thin membranes appearing thicker than they really are.

The list below describes, in simplistic form, some of the common artifacts, some of which offer useful diagnostic information.

Acoustic Shadowing

If the proportion of the beam energy attenuated at a given interface is high, little sound will be transmitted deeper than the interface, and then echoes will not be received from tissue deep to the interface. When this occurs, an acoustic shadow is produced.

Some interfaces causing such effects are:
- **Soft tissue/gas (bowel, lung)**
- **Soft tissue/bone, calcium   (ribs, calculi)**
- **Normal tissue/fibrous tissue (scars, ligaments)**

Bright Echo originating from leading edge of interface
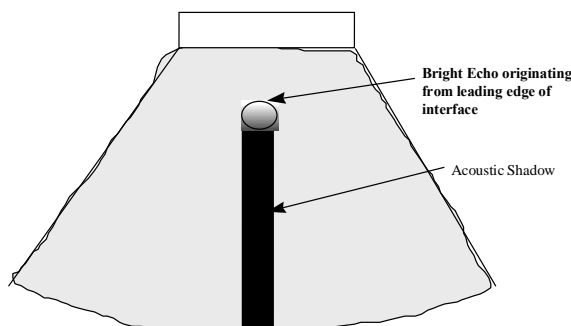
Acoustic Shadow

➲ Shadows from calcium and bone are usually echo free ("clean")

➲ Shadows from gas are often partially filled with reverberations ("dirty")

➲ Focal zone MUST be set at depth of calculus to clearly show a shadow

Acoustic Enhancement

Acoustic enhancement is an attenuation phenomenon resulting in an area of increased brightness (amplitude) in the display, relative to echoes from adjacent similar tissue.

The basis for this artifact is that fluid filled structures (cysts, gall bladder etc) attenuate sound to a much lesser degree than solid organs (liver, spleen etc). Therefore there is more sound transmitted to structures deep to the fluid filled structure and the resulting echoes from this deeper tissue are brighter than those from a similar depth in adjacent solid tissue.

Transducer

Fluid filled structure

Acoustic Enhancement
Echoes brighter than surrounding similar tissue

Edge shadow
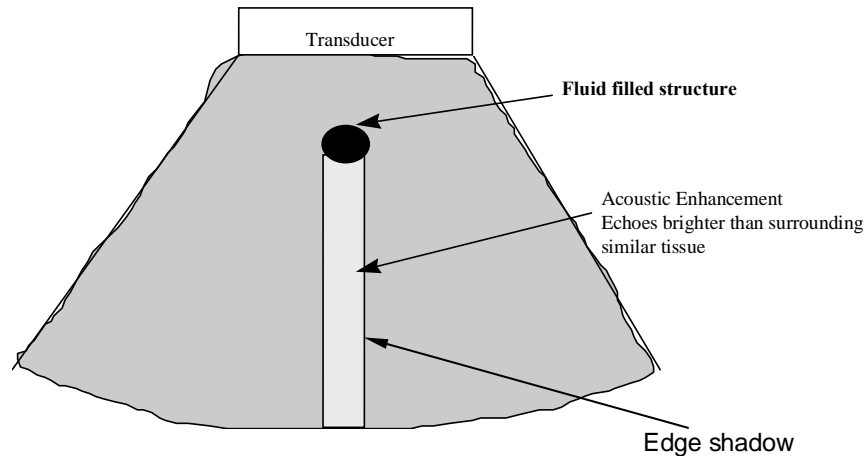
➲ **Acoustic enhancement is one of the criteria for classification of cysts (the others being smooth walls and echo free interior).**

➲ **The margins of the area of enhancement are usually sharply defined and may be parallel or divergent, depending on the transducer used.**

Edge Shadowing

Commonly seen deep to the edges of rounded structures when the velocity of sound in the rounded structure is different from that of the surrounding tissue. This results from a combination of reflection and refraction occurring at the edge of rounded structures.
(See diagram above)

Reverberation Artifact

The multiple representations in the display, of the same interface. They occur because of repeated reflections of sound between two interfaces and are usually generated by high-level mismatch interfaces when the echo amplitude is very high.

Echoes are placed in the display according to the time taken for their return to the transducer after their emission. If an echo repeatedly reflects (reverberates) between two interfaces each successive returning echo will be perceived as coming from twice the distance as the reflection before, and will be placed at twice the distance on the display.
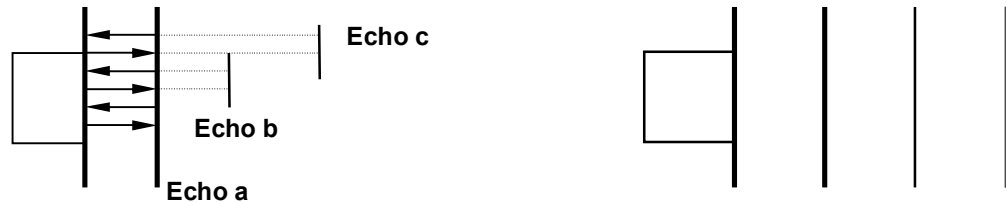
⮌Reverberation echoes are evenly spaced, because the time for each additional echo is a multiple of the time of return of the first echo.

⮌Reverberation artifacts are commonly seen in many situations, such as;
   a. **Deep to bowel gas**
   b. **Deep to subcutaneous fat and muscle layers, these are easily seen in a distended urinary bladder but may be masked by the parenchymal echogenicity of solid organs such as the liver**
   c. **Foreign bodies, multiple reverberations arising from small, superficial foreign bodies often merge and are displayed as a "comet tail" artifact**

Beam Width Effects

There are several artifacts that occur due to the reality that the ultrasound beam is not a single line of site, but that the system sees it as such.

This group of artifacts can cause much difficulty to the beginner, in that they display echoes generated from the edge of the beam, as though they were generated from the central axis of the beam. These echoes are therefore incorrectly placed in the image and can give rise to "pseudo pathology".

This group of artifacts includes:
   a. Beam width
   b. Side lobe
   c. Grating lobe
   d. Slice thickness

Common examples of this type of artifact encountered in general scanning are:

⮌Urinary bladder - "false" echoes within the bladder, arising from surrounding bowel gas, can be mistaken for pathology

⮑ **Gall bladder - "pseudo sludge", echoes within the gall bladder lumen arising from adjacent bowel or liver**

⮑ **Obstetrics - "false" echoes at the side of fetal head or femur length, can cause inaccurate measurements**

The examples above are, obviously, a minute portion of the hundreds of effects that artifactual echoes of these types can cause.

It must be emphasised that these explanations are simplistic, since the causes and effects of many of these artifacts are often complex and variable.

Velocity Artifacts

This group of artifacts occurs because various tissues have velocities different from the assumed 1540m/sec, which results in incorrect placement of echoes in the display.

The two main types in this group are;
    a. inaccurate depth calculation effects
    b. refraction effects

▷ **Examples of these are as follows:**

    a. **Low velocity lesion within the liver - causing displacement of the diaphragmatic echoes, deep to the lesion**

Liver

Low Vel Lesion

Rt Kidney

Diaphragm echo disrupted
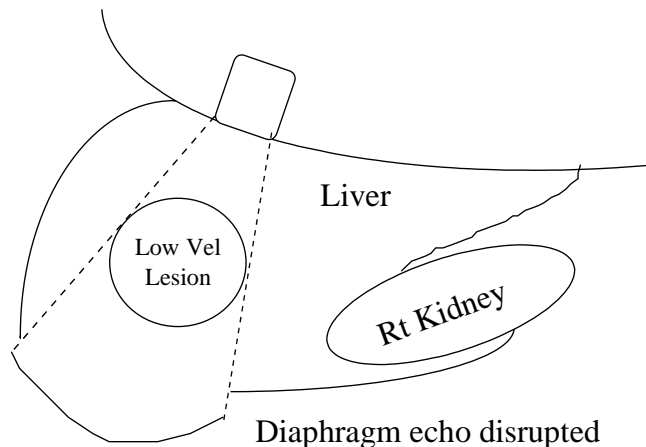
⮑ Low velocity in lesion, compared to surrounding liver

⮑ Causes echoes to take longer to return to the transducer

⮑ Therefore, the structures deep to the lesion are displayed erroneously deeper in the image

b. **Refraction effects such as "lens effect" caused by well developed rectus abdominus muscles. Results in two images of the same object, displayed side by side in the transverse plane. E.g. aorta, early gestation sac.**

Assumed line of sight

Actual path of beam (refracted)

Apparent %double
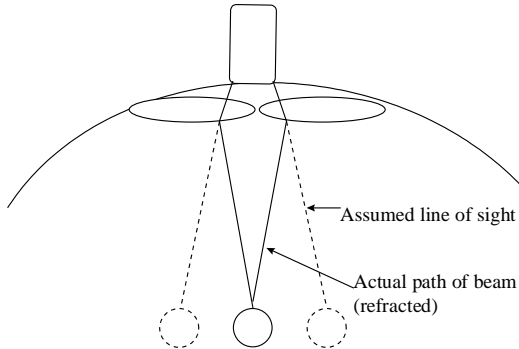
Mirror Image

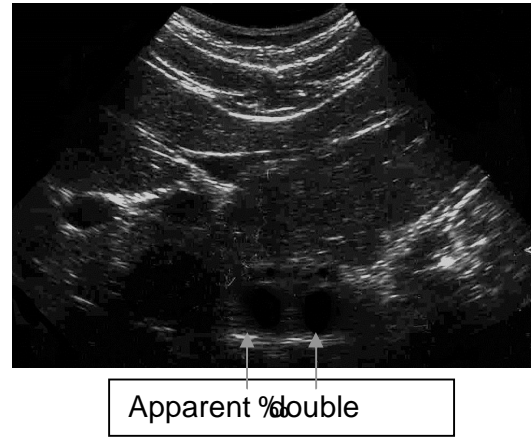Structures displayed twice, with one being a mirror of the other. These occur where there is a strong specular reflector, which acts as a mirror.

Examples of these are:

- **Diaphragm/lung interface (show mirroring of the liver texture, above the diaphragm.**
- **Bladder/rectum interface, when the rectum is gas filled (bladder wall is mirrored together with any bladder contents, such as ureterocele, catheter balloon, mass etc)**
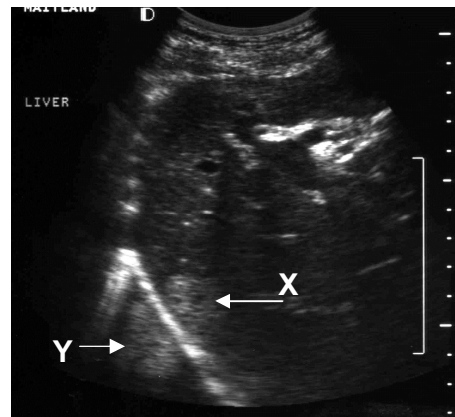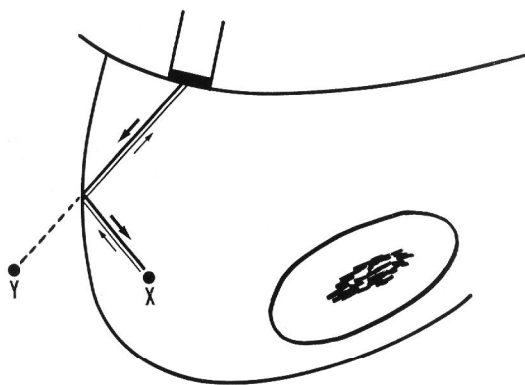
The echo from X is displayed at position Y

Multipath

The placement of echoes in the display is based on the assumption that they return directly to the transducer after reflection, but this is not always the case. Echoes can be reflected and refracted. The component of the echo that is reflected can then reflect off a second interface before returning. This results in additional path length and time; the second echo is therefore displayed deeper than the first, along the same line of sight.

Liver

Artifactual Lesion

ŏRealŏ Lesion

Rt Kidney

Operator Produced Artifacts

Inappropriate setting of gain and TGC and lack of contact can cause confusion; this is a problem common to beginners and can easily be overcome by careful manipulation of these controls.

➲ Overall Gain
It is a great temptation to turn up the gain, with the expectation that things will become clearer! However, over use of gain will merely add noise to the image.

➲ TGC
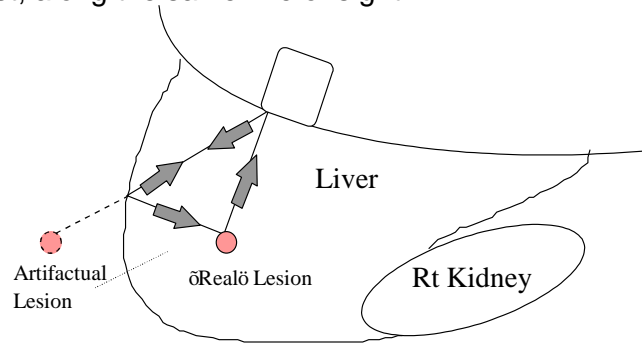**Similarly, if one of the TGC slider controls is misaligned, the slope will be uneven, possibly causing a band of bright or low level echoes within the field of view.**

➲ Contact
The most common of all artifacts produced by the beginner, is that of a lack of adequate contact between the transducer face and the skin surface. A liberal use of ultrasound gel and reasonable pressure on the transducer will overcome this.

PROBLEM SOLVING

Now that we have identified many of the common artifacts occurring in ultrasound images, we need to learn how to overcome or minimise them and to definitively recognise them as artifactual echoes and not pathology.

The simple scanning techniques required to contend with artifacts are easily acquired and can be listed as follows:

> ➲ **TURN DOWN THE GAIN**
>
> ➲ **ALTER ANGULATION OF PROBE AND WATCH THE EFFECT ON THE ARTIFACT**
>
> ➲ **SCAN IN TWO PLANES**
>
> ➲ **MOVE PATIENT**
>
> ➲ **HAVE PATIENT USE DIFFERENT BREATHING MANOEUVRES**
>
> ➲ **CHANGE TRANSDUCER FREQUENCY**

These manoeuvres, taken in order will usually be effective in minimising artifacts or at least working out what is artifactual and what is real.

At the beginning of the learning curve there is no substitute for practicing on fellow colleagues or yourself. This will enable the beginner to feel confident with machine controls and basic scanning techniques, before being confronted with a patient in an emergency situation.

SCANNING PRINCIPLES

IMAGE OPTIMISATION & KNOBOLOGY

Throughout any ultrasound examination the operator should be constantly monitoring the image quality with a view to improving it. Sometimes better scanning techniques are required and sometimes optimisation of the image by changing system controls can cause the necessary improvement in the image.

A thorough knowledge of the system controls and what effect they have upon image quality in various scanning scenarios is one of the most fundamental parts of becoming a good sonographer

TRANSMISSION POWER

Regulates the amount of energy exciting the crystal, and therefore the strength of the ultrasound beam

➲ **This should be kept to the minimum for required depth penetration, to minimise exposure**

DEPTH

The depth control alters the depth of the field of view. The number of pixels in the image area is fixed so setting the field of view also sets the number of pixels used to represent each square centimetre of the patient

➲ **Use smallest depth of field necessary to image structure of interest**

➲ **Small depth of field - superficial structures**

➲ **Large depth of field - deeper structures**

FOCUS

Modern transducers use excellent focussing technology to achieve good resolution throughout the depth of field. Ultrasound systems normally have user controlled focal zones, giving best definition at the designated depth of focus.

On some systems, it is possible to expand the focal zone by adding more than one focal point, this will allow focussing over a larger portion of the depth of field but has the trade off of slowing the frame rate.

➲ **Position focal point at area of interest.**

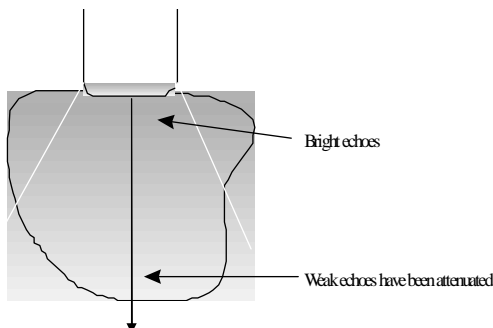➲ **Practice using more than one focal zone to watch effect on frame rate.**

TIME GAIN COMPENSATION (TGC)

As the sound beam passes through the body its strength is attenuated (diminished) by various factors as discussed above. The echoes returning from deeper structures are weaker (of lower amplitude) than those from more superficial structures. Applying gain

to returning echoes from deep structures compensates for this. The operator decides at what rate this gain will be applied by the use of the TGC controls.

NO TGC Applied
Attenuation occurs

TGC Applied
Compensates for attenuation at depth - over time

Amplitude (Brightness)

Slope

Depth/Time →

Depth/Time →

Bright echoes

Weak echoes have been attenuated

Equal amplitude ecoes from deep and superficial structures
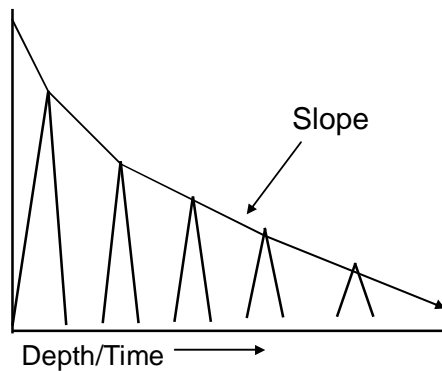
Brightness of echo decreases with depth
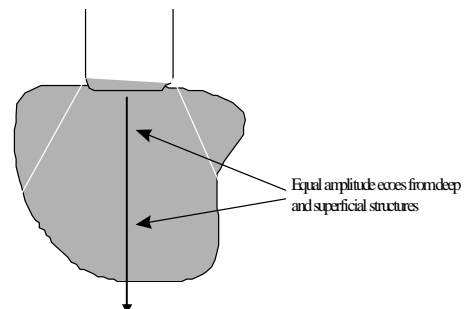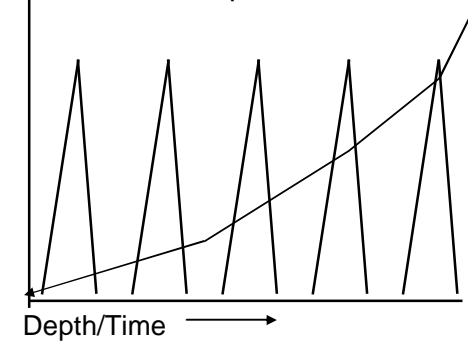
Brightness of echoes equal throughout depth of field

NO TGC Applied

TGC Applied

Ultrasound systems have user operated TGC controls which can be either slide pots, with each slider regulating one section of the image or rotary controls regulating near gain, slope and far gain. (N.B. Some systems only offer near gain and far gain control.)

➲ **TGC adjustments vary with transducer frequency**

➲ **TGC adjustments vary for each organ/system being imaged**

➲ **TGC adjustments vary patient body habitus type being imaged**

OVERALL GAIN

Amplification of the <u>Received</u> signal

➲ **Increasing the gain does not increase exposure**

➲ **Affects the whole image equally**

➲ **Undue increase in gain may obscure subtle changes in texture, or produce artifactual echoes in fluid filled structures (e.g. gall bladder, cysts etc)**

DYNAMIC RANGE

The term refers to the range of echoes from strong to weak, available to be displayed on the monitor at a particular time.

➲ **Decreasing dynamic range gives fewer grays and increases contrast**

➲ **Increasing dynamic range gives a wider range of grays and decreases contrast.**

➲ **Practice observing changes in dynamic range**

ZOOM

This control is used to magnify structures of interest and is best used in real time or write-zoom mode. Write-zoom affects image quality because the line density and pixels / unit area are increased. The consequence of these benefits is a decreased field of view. (Read-zoom is purely a magnification of a particular part of the field of view and does not affect image resolution)

➲ **E.g. easier to measure small luminal diameters**

FRAME AVERAGING

Sometimes called persistence or smoothing, this control allows the accumulation (or averaging) of echo information over two or more frames. Increasing the amount of frame averaging can enhance subtle textural differences but can cause blurring of the image and will result in a reduction in effective frame rate.

Decreasing or removing frame averaging is recommended when scanning highly mobile tissues (e.g. cardiac structures) since the effective frame rate will be higher.

CINE MEMORY

Storage of a number of previous image frames on system freeze. The number of frames retained varies with manufacturer and with memory requirement per frame stored.

The benefits of cine memory will rapidly become apparent on use of the control; the saved sequence of frames can be reviewed at will, manipulated and imaged. This can be particularly useful when scanning the very young or the very old patient, who may be uncooperative.

B-MODE COLOUR

Many modern systems allow the echoes (normally displayed in shades of gray) to be displayed in hues of colour. This can be useful in apparent enhancement of contrast resolution, since the human eye can more readily discern small changes in colour hue than in shades of gray. The choice of colour map appears to be quite subjective; most manufacturers offer a range of different maps for individual preference.

➲ **Useful practically in many different types of scan**

TERMINOLOGY

There are other controls available on ultrasound systems, many of which are different depending on manufacturersqrange of technologies and many of which are labeled using different terms but in fact have the same operational use. Two terms that are used frequently in discussion of ultrasound image production are pre and post processing.

- ✍ Pre-processing controls are those that involve manipulation of echo data before it is stored in the scan converter memory.. Those that can be changed by the operator are TGC, FOV, Frame Averaging, Dynamic Range and digitisation. Pre-processing controls are those that require re-scanning of the image with a different control setting if the image produced is not suitable or satisfactory.

- ✍ Post-processing controls are those that involve manipulation of the data after it has been stored in memory, such as greyscale maps, B-colour etc. The original data can be restored after manipulation with post-processing controls.

OTHER FACTORS AFFECTING THE IMAGE

FREQUENCY

Frequency is defined as the number of ultrasound cycles generated per unit of time.
E.g. 1MHz = 1 million cycles per second.

Many systems have transducers with the capability of being switched from one frequency to another. (%Wide bandwidth+, %Frequency domain imaging transducers+etc) The technology involved in producing these transducers varies between manufacturers and the precise physics involved can be learned from pertinent texts.

The important practical facts to note about the frequency of a transducer are how changing frequency affects image acquisition and resolution.

➥**Higher frequency - greater resolution, poorer penetration**

➥**Lower frequency - poorer resolution, greater penetration**

➥**Use the highest frequency possible to reach the necessary depth.**

e.g. Heart -	2.5-3.5 MHz
Abdo -	3.5 - 5.0 MHz
Pancreas, G.B. -	5.0 MHz
Foreign bodies etc	7.5 - 12.00 MHz

ANGLE OF INCIDENCE

The largest reflection of sound will occur at 90° to an interface; therefore the best images will result from a sound beam projected at 90° to the main area of interest.

PRACTICAL PHYSICS OF SPECTRAL & COLOUR DOPPLER

INTRODUCTION

It is essential to understand that both Spectral and Colour Doppler operate under the same physical laws. So the statement, "Colour is Doppler and Doppler is Colour", is accurate and should always be remembered. Therefore one should have a sound understanding of the principles of spectral Doppler, before progressing to colour Doppler.

It is important to understand that these two technologies share controls, terminology, limitations and artifacts and we will discuss these along the way.

THE DOPPLER EQUATION

The basis of all colour and spectral Doppler physics is encompassed in the Doppler equation.

$$\triangle f = \frac{2f_o V}{c} (\cos \varphi)$$

Where:

- $\triangle f$ = The detected Doppler Frequency Shift
- f_o = Initial Doppler Frequency
- V = Velocity of Moving Blood
- c = Speed of Sound in Tissue
- φ = The Angle between the ultrasound beam and the flow direction

All this means is that the initial (or, incident) frequency of the sound beam is different from that which is received from a moving reflector (blood) and that difference varies with

- the velocity of the moving reflector
- the angle between the ultrasound beam and the flow
- Initial (operating) frequency

DIRECTION AND THE DOPPLER EFFECT

Using the Doppler effect we can indicate the direction of the moving reflector. A practical, well-known description of the Doppler effect is that of the change in pitch of a train whistle or siren as they come towards or away from you when you are standing still. The pitch increases as the train is coming towards you and decreases as it is moving away.
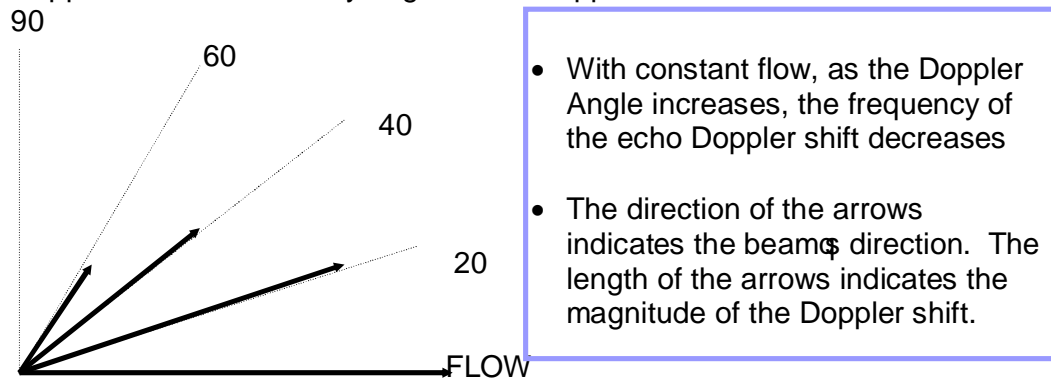
This, of course, holds true for frequency shifts detected by an ultrasound machine when the moving reflector, blood, is moving in respect to the transducer.

If the blood is moving towards the transducer then the received frequency will be higher than the initial frequency. Conversely, if the blood is moving away from the transducer, then the received frequency will be lower than the initial frequency.

However, if the blood is travelling at 90 degrees to the transducer then the received frequency will be equal to the initial frequency and therefore there will be no discernible frequency shift.

CONSEQUENCES OF DOPPLER ANGLE CHANGES

As we have seen in the Doppler equation, the dependence on the Doppler angle is in the form of a cosine. For a given flow, the greater the Doppler angle is, the lesser the Doppler shift is. A zero degree angle gives the greatest Doppler shift and at ninety degrees the Doppler shift is reduced to nil.

- With constant flow, as the Doppler Angle increases, the frequency of the echo Doppler shift decreases

- The direction of the arrows indicates the beams direction. The length of the arrows indicates the magnitude of the Doppler shift.

Flow speed calculations based on the Doppler shift measurements can only be accomplished correctly with knowledge of the Doppler angle. It is obviously important to estimate this angle as accurately as possible. This is normally accomplished by placing the Doppler angle bar, on the anatomical display, so that it is parallel to the assumed direction of flow. In straight vessels with no flow obstruction this can be assumed to be parallel to the vessel wall, however, in tortuous or obstructed vessels, this may not be the case.

Examples:

Example 1			Example 2		
Initial Frequency	-	1MHz	Initial Frequency	-	1MHz
Speed of Sound	-	1600 m/s	Speed of Sound	-	1600 m/s
Velocity of reflector	-	16m/s	Velocity of reflector	-	16m/s
Angle	**-**	**0 deg**	**Angle**	**-**	**6 0 deg**
Doppler shift	**-**	**0.02MHz**	**Doppler shift**	**-**	**0.01MHz**

CONSEQUENCES OF INITIAL DOPPLER FREQUENCY CHANGES

We must also remember the proportional relationship shown in the Doppler equation of the initial, or operating, frequency of the transducer and the Doppler shift measured by the system.

Doppler shifts measured for a given flow in a vessel, using two transducers of 3MHz and 6MHz would be different. The Doppler shift from the higher frequency transducer would be double that of the lower frequency transducer.

In the clinical situation it is then important to select the transducer frequency suitable for the particular application. In venous work, we need a higher frequency to detect the low velocities which will produce small frequency shifts.

Example 3			Example 4		
Initial Frequency	-	1MHz	Initial Frequency	-	2MHz
Speed of Sound	-	1600 m/s	Speed of Sound	-	1600 m/s
Velocity of reflector	**-**	**40m/s**	**Velocity of reflector**	**-**	**40m/s**
Angle	**-**	**0 deg**	**Angle**	**-**	**0 deg**
Doppler shift	**-**	**0 .05MHz**	**Doppler shift**	**-**	**0.10MHz**

DOPPLER INSTRUMENTATION

There are three types of Doppler equipment used for the detection of flow in the body, continuous wave and pulsed wave spectral instruments and colour mapping instruments.

CONTINUOUS WAVE DOPPLER

These instruments include a continuous wave generator and a receiver that detects the change in frequency (Doppler shift) which results from reflector motion. The information received is presented as an audible signal and a spectral display. A continuous wave instrument detects flow occurring anywhere in the overlapping region of the transmitting and receiving beams. Because there is no depth range gate, the signals can be very confusing and include a broad range of frequency shifts.

Advantage	Disadvantage
• Excellent detection of high velocity flows	• No range gate location
• No aliasing limits	• Spectral broadening

PULSED WAVE

The transducer assembly in these instruments functions both as a transmitter and a receiver. Once the magnitude and direction of the Doppler shift have been derived they are presented as both audible and spectral displays. The receiver gate (sample volume) is set by the operator at a given depth on the ultrasound display and only echoes arising from moving reflectors at that depth are displayed as a spectrum.

The upper limit of the Doppler shifts which can be detected by a pulsed Doppler system is equal to one half of the number of pulses per second (**P**ulse **R**epetition **F**requency), e.g. PRF - 1000Hz = maximum detectable Doppler shift of 500Hz. When this limit is exceeded, aliasing occurs. (This limit is sometimes known as the Nyquist Limit.) Thus higher PRFs permit higher Doppler shifts to be detected.

Advantage	Disadvantage
• Single sample volume • Accurate range location • Quantitative spectral analysis	• Aliasing - dependent on the PRF

SIGNAL PROCESSING - PUTTING IT ALL TOGETHER!

A pulsed Doppler system must receive the last echo from each pulse before it transmits the next burst. The system automatically calculates the round trip time of this burst, by knowing where the sample volume has been placed, and analyses the signals at that point in time.

Using a method called Fast Fourier Transform, which rapidly assesses all the different frequency shifts from the sample site, the system produces a spectral display on the monitor. A mathematical algorithm then completes the Doppler equation, correcting for the angle between the blood flow and the sound beam, converting the frequency shifts (KHz) to a velocity (M/s). Most modern systems will not complete this conversion if the Doppler angle is greater than 70 degrees, since the resulting velocity would have a large degree of inaccuracy.

COLOUR DOPPLER IMAGING

Colour Doppler is a display of **mean frequency**. Data from hundreds or thousands of sample volumes is processed, using a system called autocorrelation. A very simple explanation of this is that all the frequency shift data obtained from these multiple sample volumes is analysed for mean frequency, amplitude and turbulence.

This allows us to use colour Doppler information to quickly appreciate haemodynamic events occurring in the region of interest in a qualitative manner, such as -
- Existence of flow
- Direction of flow
- Existence of turbulent flow

thus providing a %road map+ for pulsed Doppler spectral analysis.

Perhaps the most important point to remember is that colour Doppler is a **QUALITATIVE** analysis of data and does not provide velocity information.
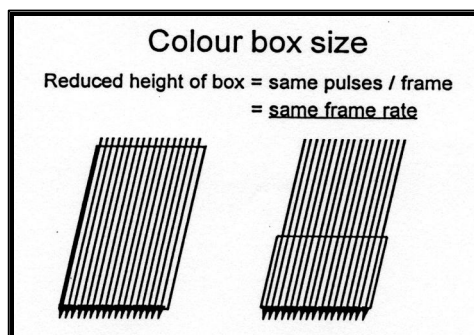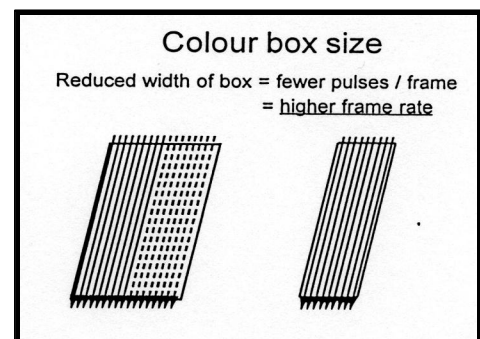

COLOUR DISPLAY

A colour wheel or colour bar is used in most equipment to display the direction, amplitude and turbulence data in a colour Doppler signal. These displays indicate direction of flow (positive or negative frequency shift) as red or blue, dependant upon allocation or encoding by the operator. The highest amplitude is displayed as the brightest colours with areas of turbulence being displayed as green (the brightest green being the most severe turbulence).

REGION OF INTEREST BOX (ROI)

The size of the colour box should be adjusted to optimise the frame rate, but still allow a suitable area of the field of view to be covered. The box may be adjusted in:
- width
- height
- depth into the patient

By reducing the width of the box, fewer colour lines of sight are required (and remember that each of these lines requires several pulses). By reducing the width, each frame can be completed in much less time therefore significantly increasing the frame rate.

Colour box size
Reduced width of box = fewer pulses / frame
= higher frame rate

Colour box size
Reduced height of box = same pulses / frame
= same frame rate

If the height of the box is reduced but the width is maintained, the same number of beam paths is still required; therefore the effect on frame rate is minimal.

If the depth of the colour box in the patient is reduced, then for each pulse sent into the patient, the computer does not have to wait as long to receive the echoes. Each line of sight can then be completed in a shorter period of time, making the entire frame completion time somewhat faster. Frame rate therefore increases.

Colour box size

Reduced depth of box = higher PRF
= higher frame rate
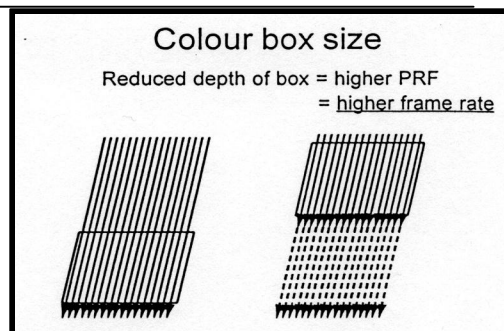
CONTROLS

Spectral and colour Doppler controls are often not properly understood by the operator because the modern systems available have excellent applications software to set the parameters for differing flow situations within the body. Most of the time, these set ups are more than adequate; however, it is necessary to understand the functions of these controls in order to optimise the system for individual, difficult cases. Ultrasound equipment manufacturers have different names for many of these controls, so it is also important to find out what terms are used in your particular system.

PULSE REPETITION FREQUENCY (PRF)

Other terms for this are flow rate or scale. This determines the sampling time that is required to process the Doppler information. Since low flow must be sampled for a longer period of time to accurately analyse the information for colour display, low PRF must be selected for these situations. Higher flows can be sampled more quickly so high PRF can be selected in these situations.

Whilst the system preset will be adequate as a beginning don't forget that changes may be necessary during an examination. E.g. Vertebral flow is often slower than common carotid flow - lowering the PRF when looking for vertebral arteries in a carotid exam can assist in speedy recognition of the vertebral artery.

FILTERS

There are numerous terms used, such as high pass filter (HPF), wall filter or thump filter. Their function is to eliminate strong, low frequency Doppler shifts arising from tissue or vessel wall motion. Another way of remembering this function is to think of them as "electronic erasers" in that the filter will eliminate or erase any frequency below that set.

Wall Filter

Filter removes these low frequency signals

Movement from walls and adjacent tissue produce low frequency shifts

Wall Filter

Filter removes these low frequency signals

eg   150 Hz
        80 Hz
        50 Hz

wall filter of 200Hz will remove all shifts

Doppler shifts (= colour signals) are produced not only from blood flow, but also from any other moving structure, including solid tissue.

Tissue motion is very slow velocity, but the Doppler signals are large amplitude compared to blood flow signals because of the large reflections from the tissue interfaces. These low velocity signals cause colour flashes to occur in the presence of moving tissue, and these can greatly degrade the quality of the colour image.
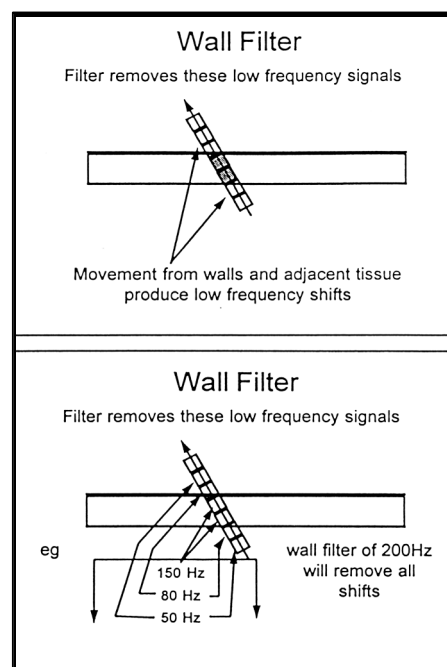
The wall filter causes the computer to completely ignore Doppler shifts
 (= velocity) below a set limit.

Because vessel walls are a major cause of these unwanted signals, the filter to remove them was called a "wall filter". However these filters will remove all frequency shifts below the selected limit, whether they are caused by wall motion, adjacent soft tissue movement or blood flow.

Although prudent use of a wall filter will produce a more diagnostic colour image, overuse (i.e. setting the cutoff limit too high), may remove significant blood flow echoes, and in extreme cases all flow representation.

When attempting to visualise slow velocity flow such as that seen in peripheral veins, the wall filter must be set very low to prevent removal of the echoes. In this case a conscious effort must be made to minimise tissue movement by careful transducer and hand movements.

In fast flow areas, such as arteries, increase the filter setting to produce an image with little colour produced over the vessel walls or adjacent tissue.

COLOUR & SPECTRAL DOPPLER INVERT

This simply allows the operator to make a negative spectral Doppler shift appear above the baseline, which can be helpful when interrogating a vessel from differing Doppler angles.

In colour, this allows arteries to be displayed in red and veins in blue, regardless of the angle of interrogation. In carotid or peripheral studies, where the vessels are mostly straight, this is fairly simple to understand and control but care must be taken in abdominal studies, where the vessels are often tortuous. The operator must have a good knowledge of vascular anatomy to avoid interpretive mistakes.

STEERING

Spectral Doppler - As we have seen, a 0 degree interrogation angle will produce the highest frequency shift. Steering allows us to place the sample volume at the best angle possible in a particular vessel and will provide directional information.
Colour Doppler - Steering the ROI will provide information on the direction of flow with respect to the transducer; however, steering also reduces the sensitivity of the system (due to refraction, scattering and aperture reduction). Therefore, when small vessel flow detection is necessary e.g. Thyroid or Testicular studies, the ROI should not be steered, allowing optimal system sensitivity.

COLOUR QUALITY

Other terms for this are packet size or length and ensemble length or sensitivity. This is simply the number of times and for how long, the system samples each line of sight in the ROI. Higher numbers will require longer sampling time and therefore reduce frame rate, but will result in a more homogeneous colour display. Lower numbers will result in higher frame rates but more pixilated colour, with a more accurate display of haemodynamic changes.

It is up to the operator to decide the quality of the colour display for each application and patient.

B-SCAN/COLOUR PRIORITY

This is a decision circuit, which allows the system to overlay colour information on areas of the greyscale image where there is no greyscale information. (That is to say, not over pixels displaying greyscale information) In large vessels, this is simple, since it is easy for the system to differentiate between the vessel and surrounding tissue. However, in vessels that are too small to be resolved on the greyscale image (testis), the moving blood detected by the Doppler cannot be written as colour in the vessels since the system does not "see" them as "black".

By reducing the priority of greyscale in the decision circuit, the operator can override the preset levels and allow this colour data to be written over pixels containing greyscale information.

ARTIFACTS

As in any other ultrasound modality, spectral and colour Doppler both have artifacts to be used or overcome. Doppler artifacts tend to be common to both colour and spectral, being displayed in different formats.

ALIASING

This is probably the most common artifact occurring in Doppler studies. The important thing to remember here is that aliasing is NOT a disease indicator. It simply means that the speed of the flow being examined is faster than half of the PRF set. Colour aliasing can be useful in certain applications to map changes in frequency shift, in a vessel with blood flowing at a constant angle with respect to the transducer.

In colour, the aliasing will be displayed as the wrong colour. Colour changing from red, through yellow and green, to blue is caused by aliasing. Colour changing from red, through black, to blue is caused by a change in direction.

In spectral Doppler, aliasing will be displayed by spectral data being wrapped around the base line. Increasing the PRF will "unwrap" the spectrum and display it in the correct direction, either above or below the baseline.

Switching to a lower Doppler frequency will also serve to decrease aliasing.

MIRROR IMAGE ARTIFACT

Presents information as flow on both sides of the baseline in spectral Doppler. This can happen when the Doppler gain is set too high or when the Doppler angle is near 90 degrees. Using electronic array probes, this artifact can become apparent when the Doppler cursor is positioned too close to the centre of the crystal.

To minimise this artifact in spectral Doppler:

- Lower Doppler gain
- Lower Doppler angle of incidence
- Reposition probe so that cursor is between the middle and edge of the crystal

Mirror imaging can occur in colour Doppler; being depicted as opposing colours in unidirectional flow or as a mirror of a vessel on the opposite side of a particularly strong reflector (Diaphragm).

- Correct angle selection will minimise colour mirroring.

ANGLE OF INCIDENCE

A vessel curving through the colour ROI makes it possible for the Doppler to interrogate unidirectional flow in two different directions, towards the transducer and away. This commonly occurs when curving vessels are interrogated at or near 90 degrees.

- Correction of the angle of incidence will eliminate this artifact

In curving or tortuous vessels changes in the angle of incidence towards 0 degrees along the vessel may display frequency shifts that appear accelerated or even aliased. This may be misinterpreted as an area of increase in velocity, but is, in fact, artifactual.

- Understanding of the physics of the effect of the angle of incidence on colour display will eliminate misinterpretation of this artifact.

PROBE MANIPULATION TECHNIQUES

Once the machine controls have been mastered, scanning skills need to be acquired in order to continue upward on the learning curve. The beginner can often find manipulation of the transducer a difficult and painful process (the fingers and wrist become locked and sore!!).

Recognising the orientation of the transducer is probably the most important factor to begin with. Most transducer manufacturers provide an orientation mark, usually a small raised bump or light on one side of the transducer handle. In general, this mark should be towards the patient's head for sagittal sections and towards the patient's right for transverse sections.

The transducer should be used as an extension of your hand and visualised as a torch, with the ultrasound beam as the light emanating from the torch. The anatomy must be visualised as three-dimensional and not just the two-dimensional view seen on the monitor, point the transducer at the organ you wish to image, remembering the three dimensions.

The probe should be held with the operator's fingers close to the transducer face and with the little finger resting on the patient's skin, so that maximum control is exerted. In this way, the transducer will not slide off the scan plane needed and can be kept steady on the patient's skin, at the correct angle.

➲ **Orientate the transducer correctly**

➲ **Remember the body is three-dimensional anatomy**

➲ **Use the ultrasound beam as a torch, pointing at what you need to see**

➲ **Hold the probe so that the fingers control the transducer face properly**

SCANNING TECHNIQUES

1. Work Station

Operator comfort is an important factor in successful image acquisition. The operator should be positioned so that the forearm and hand are gently resting on the patient, with the spine erect and eyes at the level of the system monitor.

➲ **Scanning can be undertaken either seated (adjustable chair with wheels) or standing, depending on the height of the operator.**

➲ **If the patient trolley is adjustable, adjust the height to suit the operator.**

Whilst this is not always possible in venous scans, the benefits accrued in operator comfort and improved scans are well worth the few moments it takes to set up a reasonably ergonomic workstation for each scanning situation.

2. ULTRASOUND GEL

For a variety of reasons, it is important to use sufficient gel over the proposed scanning area. A common mistake made by beginners is to use insufficient gel, thereby making poor contact with the skin surface (producing artifacts and poor imaging) and reducing the mobility of the transducer face over the skin (producing patient discomfort and operator frustration).

➲ **Too much gel is far better than not enough!**

➲ **Spread out the gel over the proposed scanning area, before beginning your survey scan**

3. SURVEY SCAN

Always perform a swift survey of the area in question, before focussing on the point of clinical question. This will serve to orient you to the anatomy under review and identify good sonographic "windows".

➲ **Perform a survey scan first**

SONOGRAPHIC "WINDOWS"

A sonographic window can be described as a path for the sound beam into the body, to reach the organ in question, avoiding areas of bone, bowel, or lung. An example of this is:

➲ **Using muscle bellies to view veins from different approaches & to avoid shadowing from ligaments or fascial planes.**

A large part of the sonographic art is an intuitive grasp of finding good sonographic windows. Visualising the body in three dimensions and applying logical thought to the best approach for a particular organ, vessel or lesion will quickly lead to this intuitive grasp.

5. TRANSDUCER MOVEMENT

Transducer movements should be smooth free flowing, allowing proper time for the operator to view the real time image on the monitor without constant checking of transducer position. When the area of interest has been reviewed and the sonographic window decided upon, the transducer movements should be slowed and small movements made in one plane at a time, otherwise it easy to become confused and lose sight of the organ, vessel or lesion in question.

Transducer movements are;
1. Rotate
2. Fan (with long axis of transducer)
3. Heel / Toe (H=non-marker / T=marker)
4. Slide (towards or away from marker)

➲ **Make one movement at a time, with logical forethought**

➲ **Slow movements down when focusing on area of interest**

GRADED COMPRESSION

The use of firm pressure on the transducer can significantly improve images of deeper structure by several means:
a. decreasing the distance of the target to the transducer, thus bringing the target into the main focal area of the beam or
b. allowing the use of higher frequency transducers or
c. pushing loops of gas filled bowel out of the field
d. improving skin/transducer contact
e. minimising some artifacts

The emphasis should be on gradual, graded compression - since this is well tolerated by most patients, even those experiencing significant pain.

➲ **Use graded compression, increasing and decreasing pressure gradually**

8. MULTIPLE SCAN PLANES

Structures should always be viewed in at least two different scanning planes, with longitudinal and transverse sections being the minimum. It is preferable to view from two different approaches as well, e.g. anterior & coronal. This can easily resolve confusion concerning the nature of some structures e.g. cyst versus tube, aorta versus inferior vena cava.

➲ **Always view structures from at least two different approaches and in two different planes.**

DYNAMIC TECHNIQUES

Holding the transducer steady and watching muscles & tendons move in and out of the field as the patient gently moves can often elicit a large amount of information as to the best window to use.

THE TRANSDUCER AS A PALPATION TOOL

Using the face of the transducer to gently palpate an area of pain will often elicit clinically useful information, e.g. acute pain on direct palpation of superficial thrombophlebitis. The benefit accrued from this is being able to view the area being palpated to be certain of cause of pain.

PROTOCOLS AS A USEFUL TOOL

Departmental protocols are an extremely useful tool during the learning curve and should be used regularly until they are second nature. A simple way to enhance memory of different scanning techniques is to make up your own protocols using your own words.